External fixators for pelvic fractures

Comparison of the stiffness of current systems

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ABSTRACT We evaluated the stiffness of external fixation (EF) systems with a reproducible, standardized human pelvic replica of aluminum and perspex in which a type C pelvic ring injury was created. 12 EF systems were analyzed in 2 situations that necessarily occur during a walking cycle. Endpoints were defined as 15 mm of dislocation or tolerance of the maximum load in each situation.

In the no weightbearing situation, all except 2 fixators failed; in the weightbearing situation, all fixators failed. Single bar systems performed better than frame configurations.

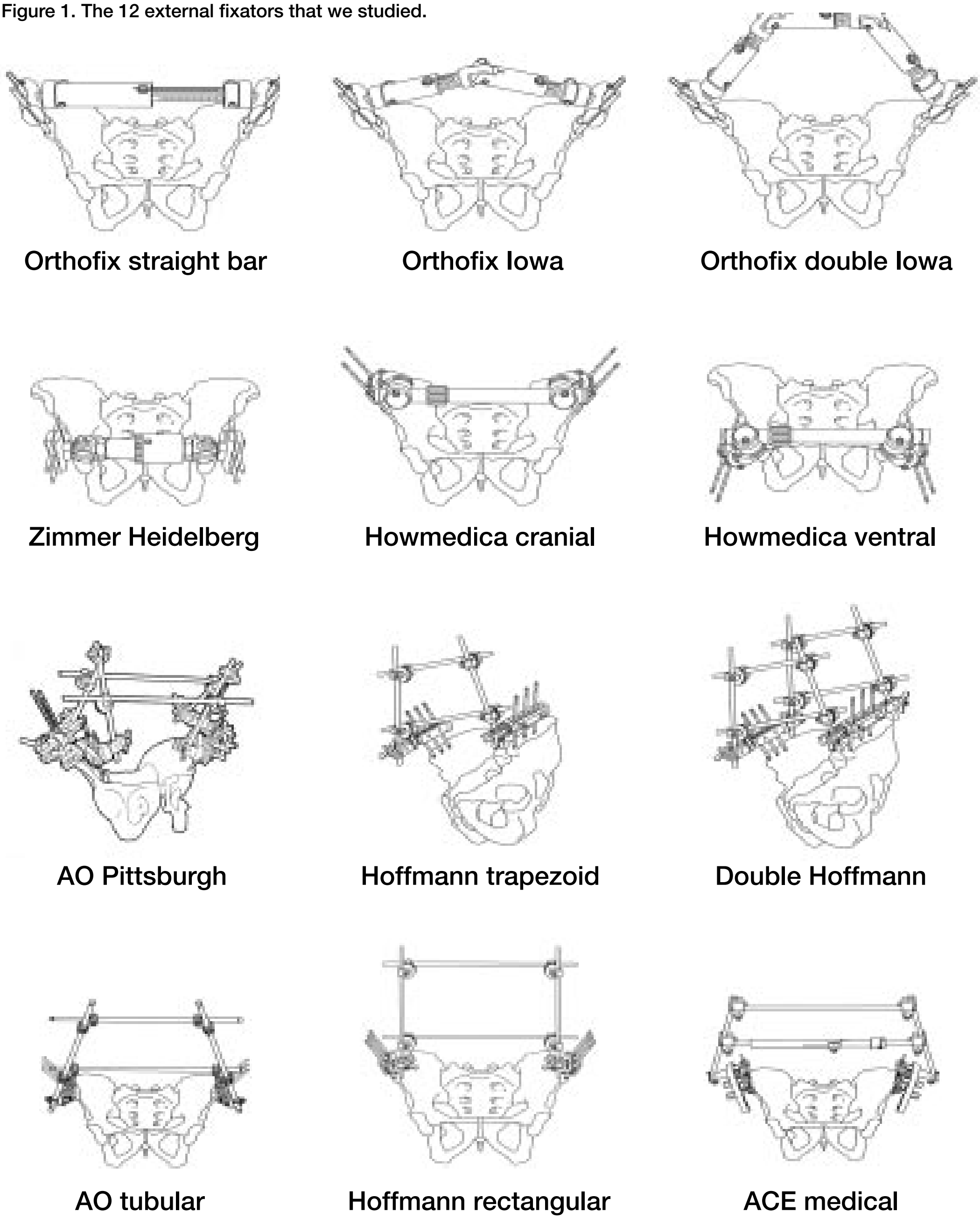
Stability provided by any external fixator is low, and in the case of a type C pelvic ring injury, it is insufficient for patient mobilization and weightbearing. Single bar systems provide more stability than frames.

Several studies on external fixation of pelvic injuries have been published since the 1970s (Slätis and Karaharju 1975, Poka and Libby 1996). Initially, large heavy frames with relatively small pin diameters were used. Nowadays, designs are more compact, easier to adjust, more comfortable for the patient and the pins are thicker. Clinical applications comprise the resuscitation phase (Gylling et al. 1985), initial fracture stabilization (Matta and Saucedo 1989) and sometimes definitive fixation (Tile 1988). Although extensive clinical experience has been obtained (Tile 1988), knowledge about underlying mechanical principles remains limited (Behrens 1989a,b). Reduction of the fracture and preservation of stability of the fixation are of chief importance. These mechanical principles can be expressed in terms of pressure between fracture surfaces and resistance against (re-)dislocation of the fracture—e.g., during partial weightbearing in the (re-)convalescence period. Various types of commercially available external fixation systems are used to satisfy the mechanical requirements. For the trauma surgeon, however, the question remains as to which system is better. We tried to determine which system provides more stability and stiffness. For this purpose, we measured the 3-dimensional load-deformation behavior of 12 systems fixed to a model of a pelvis with simulated fractures.

Material and methods

Current external fixation systems were obtained from our clinic or were provided by the respective companies. With these external fixators, various configurations were constructed according to descriptions in the literature or as described by the manufacturer. In total, 12 configurations (Figure 1) were mounted on a model of a pelvis and were tested by the use of a load. A distinction can be made between frames, consisting of multiple slender rods and smaller pins, such as the AO Tubular Slätis frame versus single bars, consisting of solid, overdimensioned bars and thicker pins, such as the Orthofix Iowa (Figure 2). The pelvis model was positioned in anterior tilt with an inclination of

Figure 1. The 12 external fixators that we studied.



45°of the endplate of S1 with respect to the vertical to simulate erect stance. Frames were mounted at an angle of 70° between the fixator and the vertical plane, according to Slätis (Pennal et al. 1980). The numbers of pins were based on descriptions in the literature, or on the descriptions provided by the manufacturers (Tables 1 and 2). In Figure 3, the positions of the pins are indicated. In the ventral pin position, the first pin is inserted 1 cm cranial to the anterior inferior iliac spine and the next pin is inserted 2 cm more cranial to the first one. In the superior pin position, the first pin is placed on top of the pelvic rim 1 cm posterior to the anterior superior iliac spine, the next pin(s) is (are) inserted

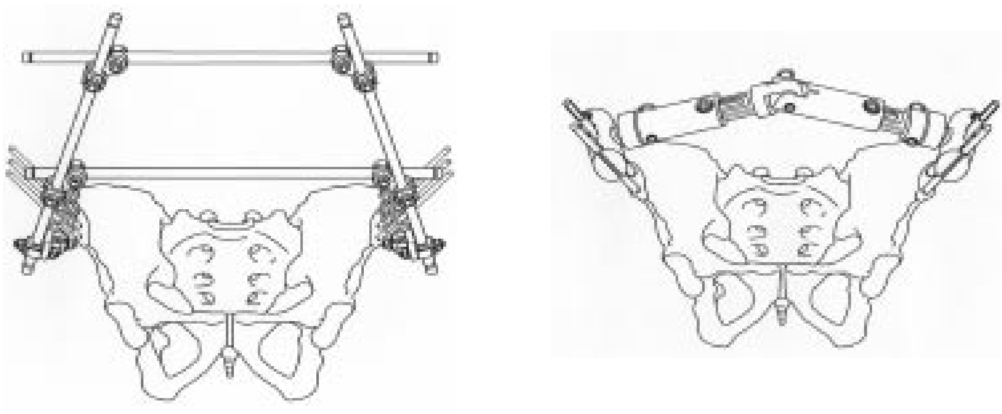


Figure 2. An example of a frame configuration (AO Slätis /left) and a single bar system (Orthofix/right).

2 cm dorsally to the previous pin. Insertion depths of 50 mm in predrilled holes in the perspex hip bone were standard. In all single bar systems, 2 pins on each hemipelvis were used, while for the frames, 3 pins on each hemipelvis were used, except for the Pittsburgh configuration, which prescribes 2 clusters of 2 pins at each hemipelvis. The distance from perspex to clamp on the fixator pins was standardized. In the ventral pin position, this distance was 6 cm. In the cranial position, this distance was 3 cm.

The pelvis model was mounted on a specially-developed loading apparatus, which had been used in a study on a human pelvis specimen (Figure 4) (Vleeming et al. 1989). The pelvic model was made of perspex and aluminum, which ensured identical geometry in each test and gives a reproducible method of pin fixation. At the sacroiliac joint and the pubic symphysis, the model was interrupted to create an unstable pelvic ring disruption (Tile/AO type C1). One of the hemipelvises was connected to the rigid base of the loading apparatus. Each

Table 1. Frame fixators

Fixator	Position of pins (cranial/ventral)	Number of pins/side	No weightbearing Fmax (N)	Weightbearing Fmax (N)
Ace medical	Cranial	3	10	10
AO tubular	Cranial	3	34	37
Hoffmann trapezoid	Cranial	3	37	39
Hoffmann rectangular	Cranial	3	45	43
AO tubular Pittsburgh	Cranial+Ventral	4	44	45
Hoffmann double	Cranial	3	65	67

Table 2. Single bar fixators

Fixator	Position of pins (cranial/ventral)	Number of pins/side	No weightbearing Fmax (N)	Weightbearing Fmax (N)
Orthofix straight bar	Ventral	2	63	74
Howmedica cranial	Cranial	2	73	77
Howmedica ventral	Ventral	2	60	91
Orthofix double Iowa	Cranial	2	143	153
Zimmer	Ventral	2	116	159
Orthofix Iowa	Cranial	2	142	161

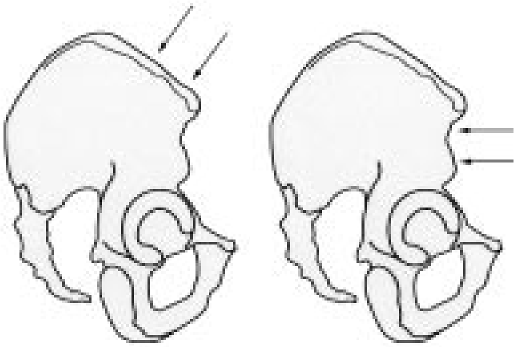


Figure 3. Lateral view of cranial (left) and ventral (right) pin positions .

time before an external fixator was applied, the pelvic ring was restored and both the hemipelvises were rigidly connected by a special construction to guarantee an identical starting position in every test. At the start of each test, this connection was removed. The "free" hemipelvis was loaded with a traction motor device.

Two loading situations (A and B) were used (Figure 5). The first situation (A) simulates weightbearing on the leg of the uninjured side. In vivo in this situation, the "free" hemipelvis bears the load of the free hanging lower extremity. This situation was simulated by means of a vertical pulling force acting at the center of the acetabulum. In this study, this is called the no weightbearing situation. The other situation (B) simulates the weightbearing on the leg of the injured hemipelvis. Here, the "free" hemipelvis has to bear the full bodyweight minus

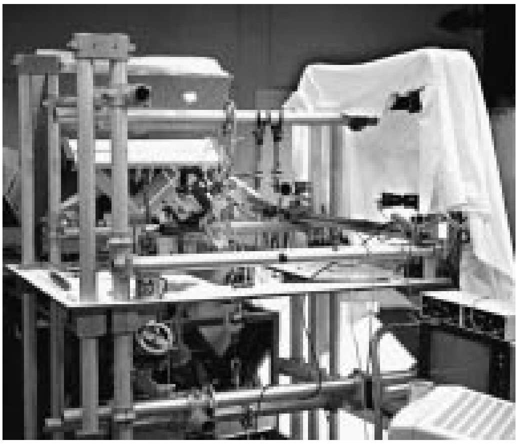


Figure 4. Loading apparatus.

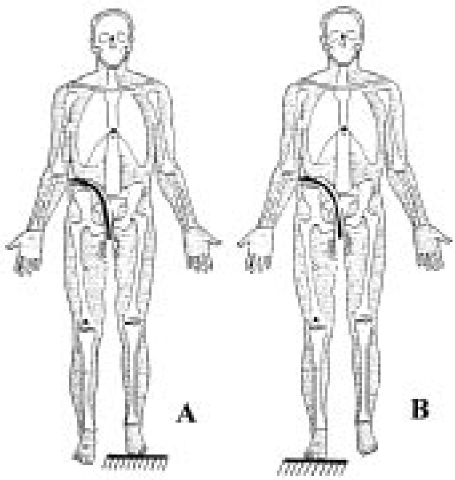


Figure 5. Loading situations: A) no weight bearing and B) weight bearring.

the load of the weightbearing leg. The latter situation, called the weightbearing situation, was simulated by means of a vertical pulling force acting at a point just ventral to L5 and 15 mm lateral from the midline at the side of the hanging leg. In both situations, the force was gradually increased, while the dislocation of the surfaces of the sacroiliac joint was monitored in relation to the force applied. The force was increased until the dislocation exceeded 15 mm or until the chosen maximum force was applied. In the first test (A) the maximum load was defined as the weight of the lower extremity (140 N); in the second test (B), this load was defined as the total body weight minus the weight of the weightbearing leg (560 N). Both of these loads are an estimate of the loads in a person of 70 kg (700 N).

The displacement between the joint surfaces was recorded with the help of a video system. This consists of a minicomputer that digitizes and analyzes video images (Keemink et al. 1991). 4 reflecting markers were placed on each hemipelvis. These were registered with 2 cameras having an infrared light source. The computer can recognize these reflections. By combining the information from both camera images, the coordinates of each marker are calculated in 3 dimensions with the help of the Direct Linear Transformation Method (Abdel-Aziz and Karara 1971). With this technique, 2 (or more) cameras are directed toward a calibration body with markers at known positions. For both cameras, the image coordinates of each

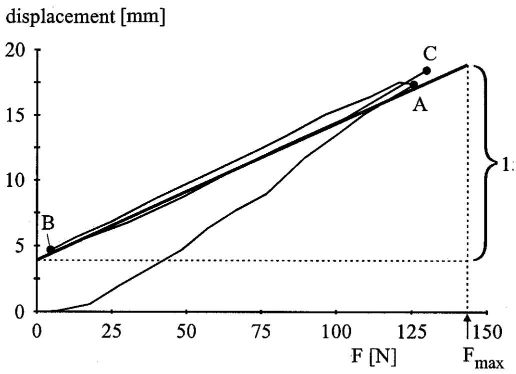


Figure 6. Example of a load-displacement curve.

marker are registered. From this registration, 11 DLT parameters are calculated for both cameras. These parameters define the relation between the spatial coordinates and the image coordinates. After this, the calibration body is replaced by the pelvis model. With the help of the DLT parameters, the spatial coordinates of the markers on the model are calculated from the image coordinates of the 2 cameras. The position of the joint surfaces in relation to the markers is known. Thus from the position of the markers, the position of the joint surfaces can be calculated.

A typical loading displacement curve is shown in Figure 6. Loading was increased incrementally until 15 mm dislocation or until the maximum load was reached, the trajectory '0-A' on the graph. During the first loading cycle, the increasing force causes the thread of the pins to cut further into the perspex. Because of this, no system completely returned to the initial situation when the load was released (see point B in Figure 6). This phenomenon, that we called 'settling', occurs only once so long as the direction of the load is not inverted. Such settling did not occur in this study, but we were interested only in the stiffness properties of the fixator system and not in the behavior of the pin in the perspex. From this situation, another loading cycle was performed, again up to 15 mm dislocation (the trajectory 'B-C' in Figure 6). This second curve reproduced well, and was used as a load-displacement characteristic. On this curve, a linear fit was performed (the thick line in Figure 6). The slope of this line is a measure of the stiffness of the fixator. In the present article, this stiffness is expressed as the force Fmax needed to produce a dislocation of the maximal allowed dislocation of 15 mm (Figure 6). In the following text, the Fmax is referred to as 'failure load'. This force is considered as a measure of the performance of the fixator systems. The performance of the frames as a group is compared with the performance of the single bar systems as a group, using the Mann-Whitney test for nonparametric distribution in 2 independent groups. A p-value less than 0.05 was considered to be statistically significant.

Results

In the no weightbearing situation, the failure load of the external fixation systems ranged from 10 N for the ACE Medical configuration to 143 N for the Orthofix Double Iowa system. The failure load of the frames ranged from 10 N for the ACE Medical configuration to 65 N for the Hoffmann Double Trapezoid configuration. The failure load of the single bar systems ranged from 60 N for the Howmedica Monotube to 143 N for the Orthofix Double Iowa system.

In the weightbearing situation, the failure load of the external fixator systems ranged from 10 N for the ACE Medical configuration to 161 N for the Orthofix Iowa system. Failure load of the frames ranged from 10 N for the ACE Medical configuration to 67 N for the Hoffmann Double Trapezoid configuration. Failure load of the single bar systems ranged from 74 N for the Orthofix Straight Bar to 161 N for the Orthofix Iowa system.

All single bar systems (Table 2) showed less displacement than the frames (Table 1). In the simulated no weightbearing situation, 10 of 12 external fixators dislocated 15 mm before the maximum load of 140 N was reached.

In the weightbearing situation, none of the external fixators resisted the maximum load of 560 N.

Discussion

When simulating standing on the injured and uninjured leg, we found typical load-displacement curves. Stiffness of all external fixator systems was low. As a group all frame configurations performed less well than the single bar systems.

In the no weightbearing situation, doubling of the Hoffmann Trapezoid frame configuration resulted in a failure load of 65 N, which places the Double A frame in the same range as the Howmedica and Orthofix straight fixators. The best results were found using the single bar systems in cranial position, with a failure load of 143 N for the Orthofix Double IOWA fixator.

The results were very similar in the weightbearing situation. With one exception, all single bar systems showed higher failure loads than the frame configurations. Again doubling the trapezoid frame almost doubled the stiffness (Hoffmann Double A). Construction of a double parallel frame increases stability of the system. However, the construction of such a frame is time-consuming, which makes it less suitable in the acute situation. Moreover, the frame is very bulky, and the same pelvic stability can be obtained with simpler constructions, which are more comfortable for the patients.

The weakest frame fixator was the preassembled ACE Medical Slätis frame, partly because of the weaker mechanical properties of the (prescribed) titanium fixator pins. Neither a trapezoid or rectangular configuration nor the choice of material (AO tubular rods or Hoffmann solid rods) seems to make a difference.

The single bars were stiffer than the frame configurations. This remarkable difference can be ascribed to:

a) The use of 6 mm pins instead of 5 mm ones (since 2 pins of 6 mm give 15% more stiffness than 3 pins of 5 mm);

b) The use of stiffer bars instead of the more slender rods of the frames.

Many comparative studies have used fresh or embalmed human specimens (Pennal et al. 1980, Vecsei 1988, Pohlemann et al. 1994), which includes an inconstant factor because of interindividual differences. To avoid such errors, we used a model of perspex and aluminum, which implies constant stiffness of the model (Broekhuizen et al. 1990). The geometry of the pelvis was based on the human pelvis, because lever arms of fixator parts as regards points of load application, and the location of the fractures, are essential for such calculations.

The dislocation was always more in the posterior part of the pelvic ring than in the pubic symphysis. Therefore, we had the dislocation at the SI joint determine the maximum applied load. Unlike other authors (Bell et al. 1988, Rieger et al. 1996), we recorded the total amount of posterior dislocation in 3 dimensions. In every case, however, at least 90% of the total displacement occurred in the vertical axis. This finding shows that when loading occurs in a vertical direction, displacement measurements in the vertical direction alone will be reliable.

Moreover, unlike other studies—e.g., that of Rieger et al. (1996)—the amount of contact and friction at the sacroiliac joint also differed. Rieger uses an unstable pelvic ring injury, but assumed anatomical reduction with full contact, but they did not apply compression. They therefore created an optimal situation for external fixation of this type of fracture. However, as was shown by Bucholz (1981) in his postmortem evaluation of pelvic ring injuries, closed anatomical reduction can not be achieved in many cases. For this reason, the amount of contact, friction and compression at the sacroiliac joint varies.

In our test model, no contact existed between the fracture parts—i.e., both hemipelvises were completely separated. In practice, various types of pelvic fractures or luxations have different resistance to motion, which depends on the amount of reduction of fracture and on the compression at the fracture site produced by the fixator. To avoid these uncertain factors, we eliminated contact at the fracture or luxation site during the experiments. Therefore, the external fixator itself was the only variable in the test.

In summary, we found that a modern single bar external fixator is more stable than frame systems. External fixation in an unstable pelvic fracture is usually insufficient to produce full stability.

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None competing interests declared.

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